
# A GaAs Pixel detector based digital mammographic system: performances and imaging test results

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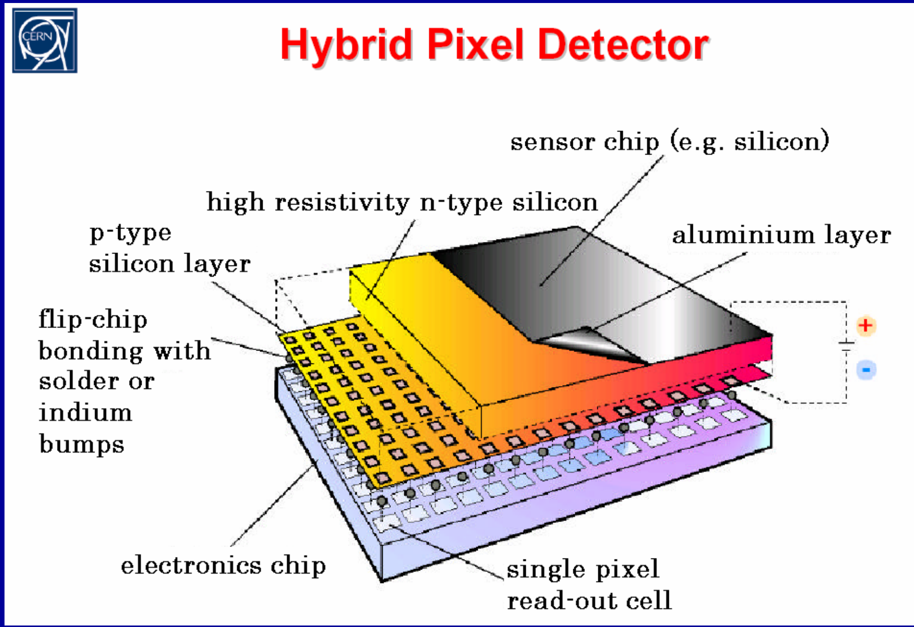
# A technological know how transfer project

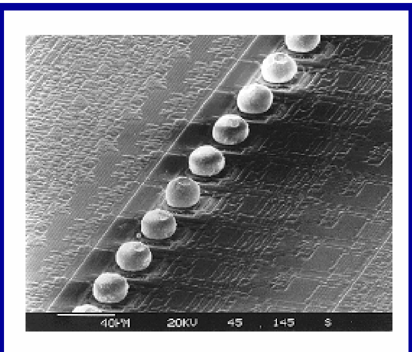
Researchers have spent many man-years in the attempt to adapt detector and read-out technologies, originally developed in the field of High Energy Physics, to the domain of biomedical apparatuses.

These studies have been integrated in a 3 years project, funded by the Italian Government through the law 46/82 (art.10): the Integrated Mammographic Imaging (IMI) project that started in June 2000.

- Research lines:
  - An advanced gamma camera for scintimammography
  - GaAs detectors and bump bonding techniques
  - A prototype of a mammographic head and related developments
  - A high intensity tunable quasi-monochromatic X-ray source

# Hybrid pixel detectors for single photon counting
## CMOS *technology*

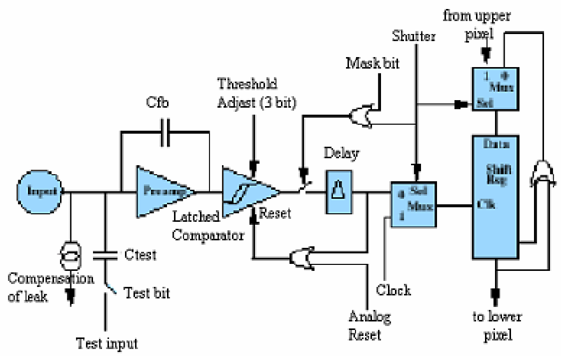


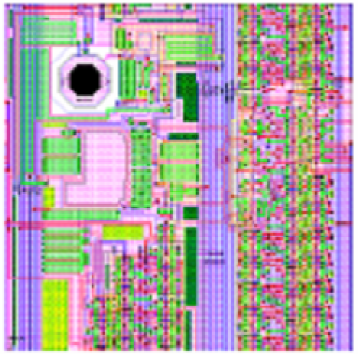



Bump-bonding: an industrial technology still posing yield problems

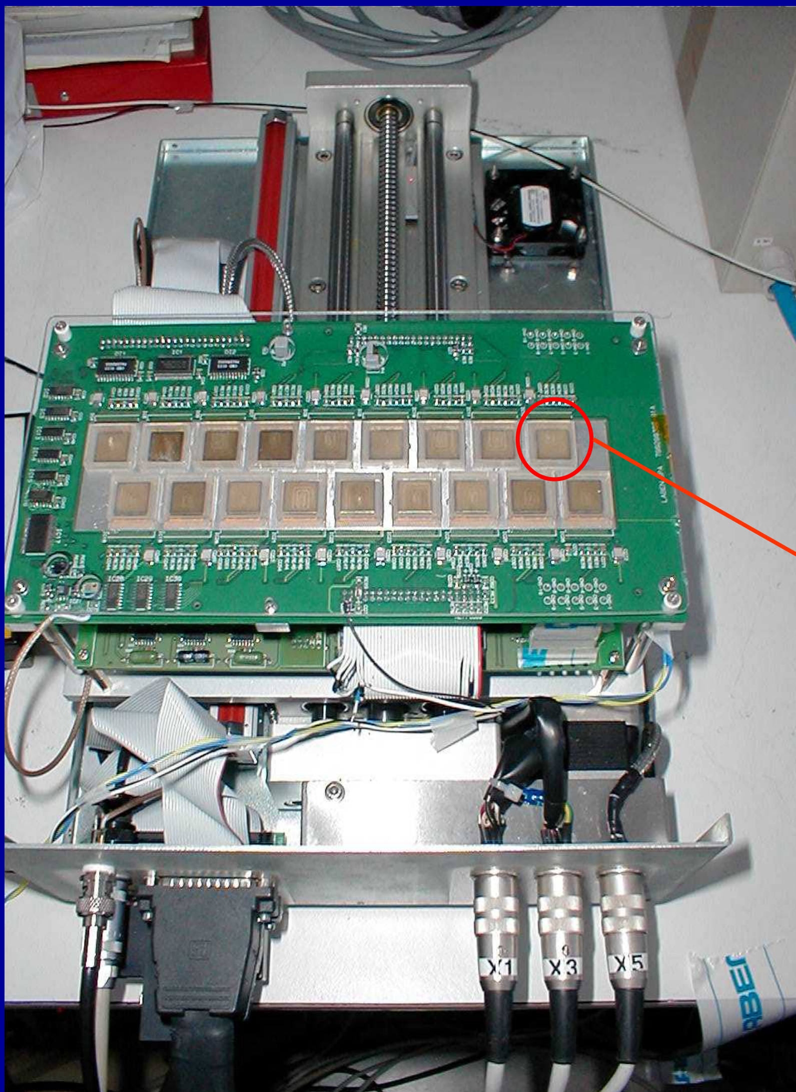
**Medipix1** (170 μm pitch) 1 μm SACMOS (M. Campbell et al.,1998)

400 transistors/cell
1.6 M transistors/chip

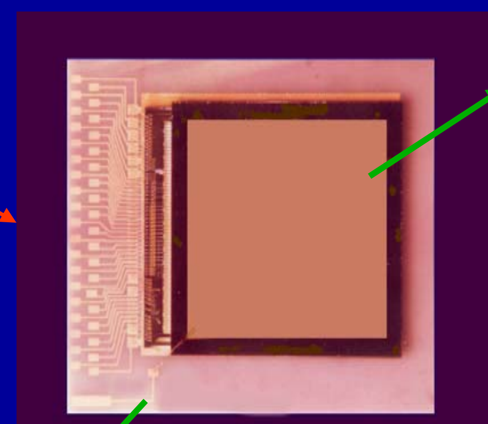


# The Detection head

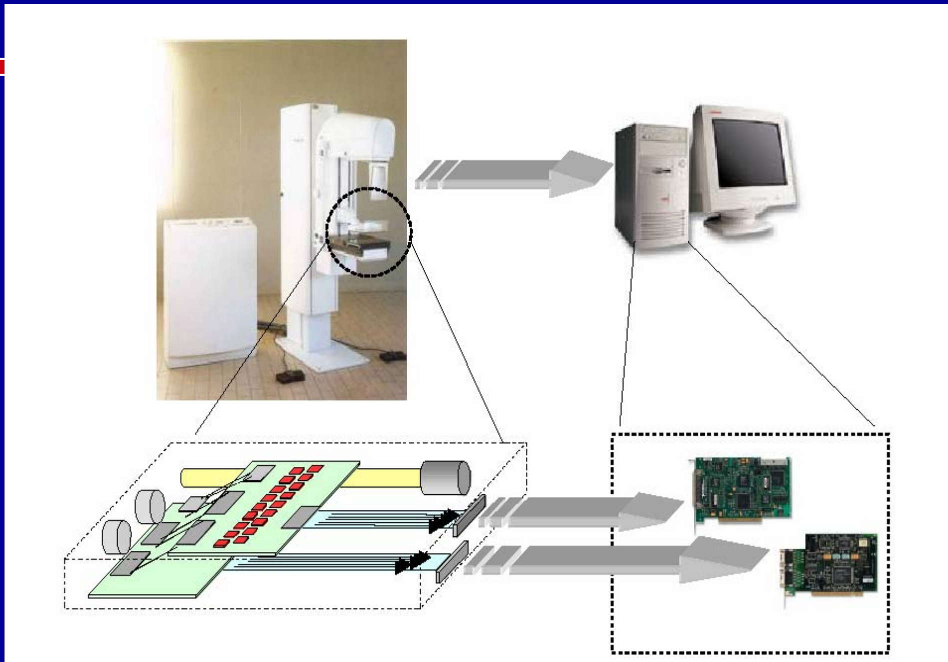


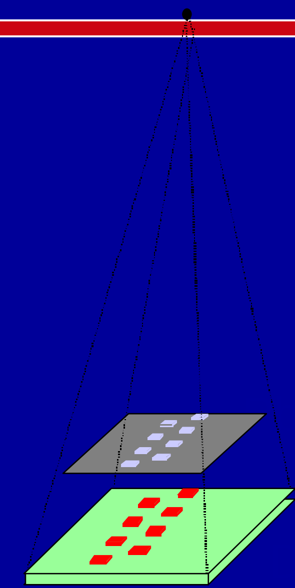
- The Detection Unit
  - The assemblies have been produced and bump bonded by Alenia Marconi Systems (Roma)
  - Each detection unit has been mounted in a protective case.



Aluminum nitride (AlN) substrate,
LEXAN cover on top (not shown)

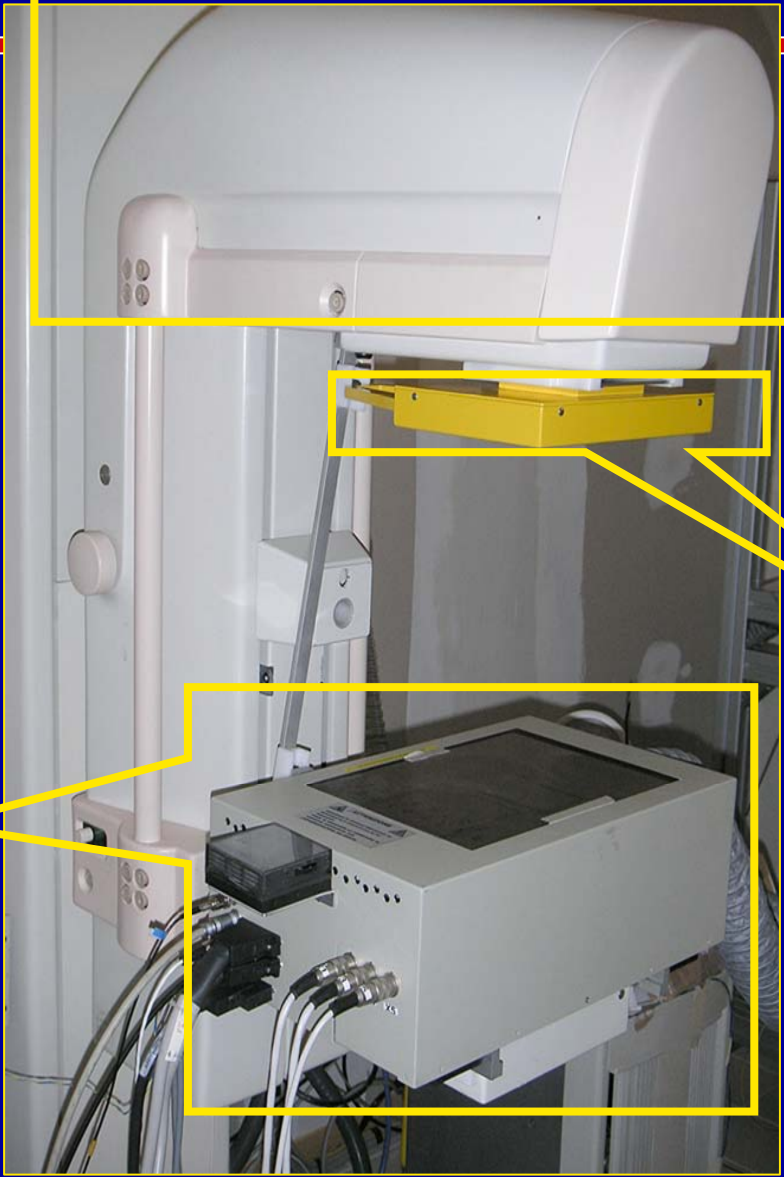
# The IMI demonstrator





- Mosaic geometry with staggered rows of nine assemblies each one,
- Scanning across the exposure field by means of a stepper motor.
- The read-out system MMRS (Multichip Medipix Readout System) (LABEN)
  - custom designed chipboard
  - interface board (IB), connected to the chipboard, implements all the logic functions of the MMRS
  - PCI board commercial board (NI 6533) that interfaces the IB to the PC.
- The BIAS Board (CAEN) provides the analog and digital biases of the chips.
- The X ray tube (Gilardoni) is the clinical mammographer SYLVIA adapted to host the collimator and the detection head.
- A controller PC handles the whole image acquisition process by means of a software program written in LabWindows CVI.

# The IMI Demonstrator



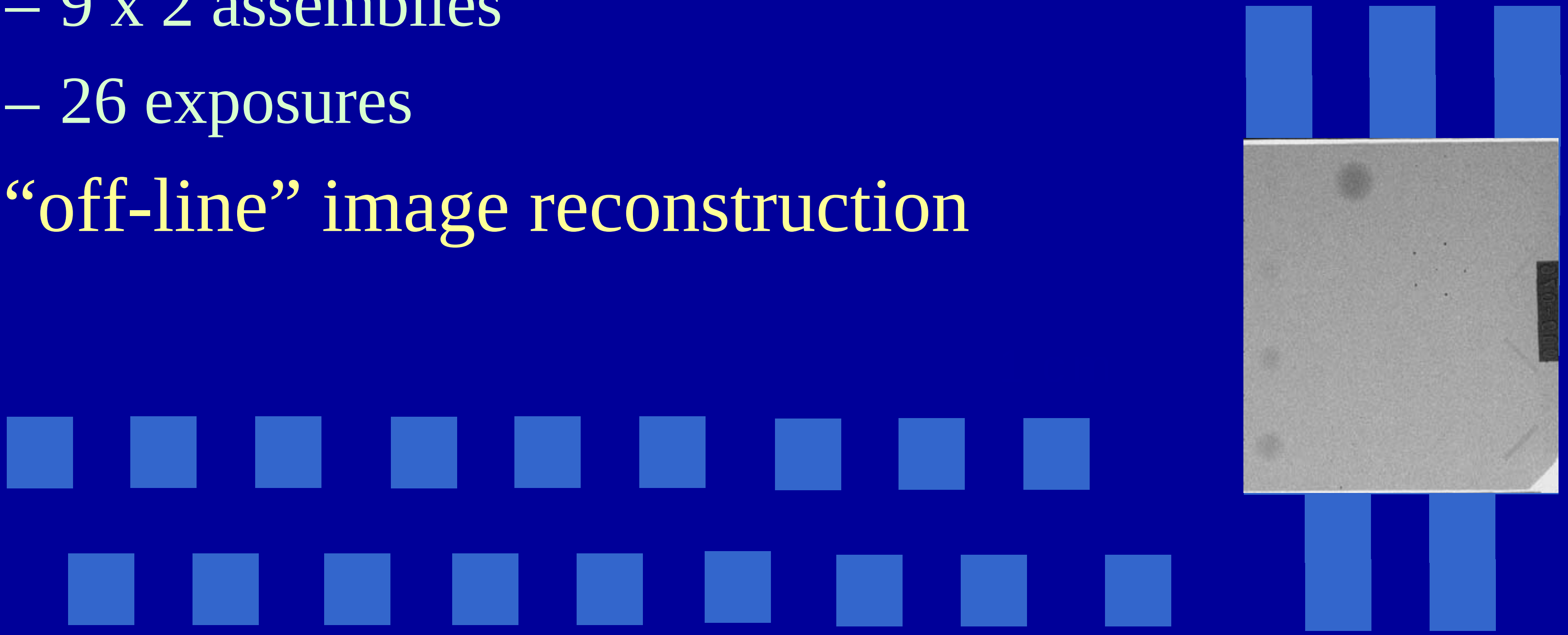
X ray tube

LEAD COLLIMATOR

Mammographic Head

# Scanning simulation

- 18 x 24 cm2 exposure field
  - 1D scanning
  - 9 x 2 assemblies
  - 26 exposures
- "off-line" image reconstruction

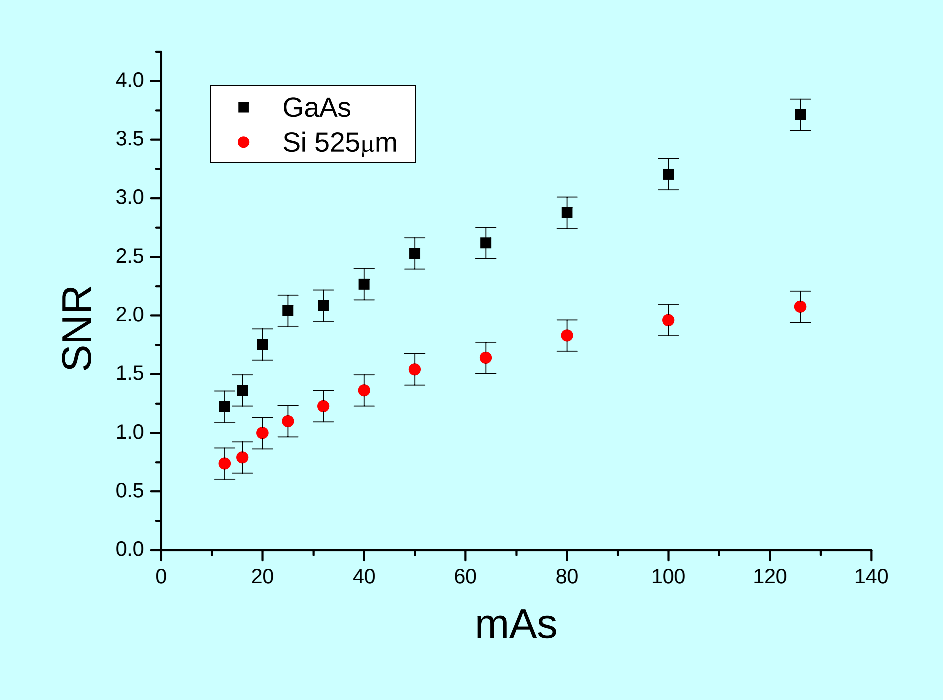


# GaAs Detectors

- Advantages:
  - High atomic number, high density, high detection efficiency
- Problems:
  - Material (SI LEC)
    - deep levels (traps)
    - active thickness, charge collection efficiency
  - Contacts
    - High voltage
- Goals:
  - High bias voltage (> 350 V)
  - Low leakage current (lower than 5 $\mu A/cm^2$)
  - High charge collection efficiency (> 75 % at the operating bias 350V)
  - Reliability

# Comparison between GaAs and Si

- Evaluation of the SNR as a function of the incident flux for the 2 mm tumor mass of the RMI phantom
  - Detectors: 200 μm GaAs and 525 μm Si
  - Tube settings: 40 kV and 16 -> 125 mAs

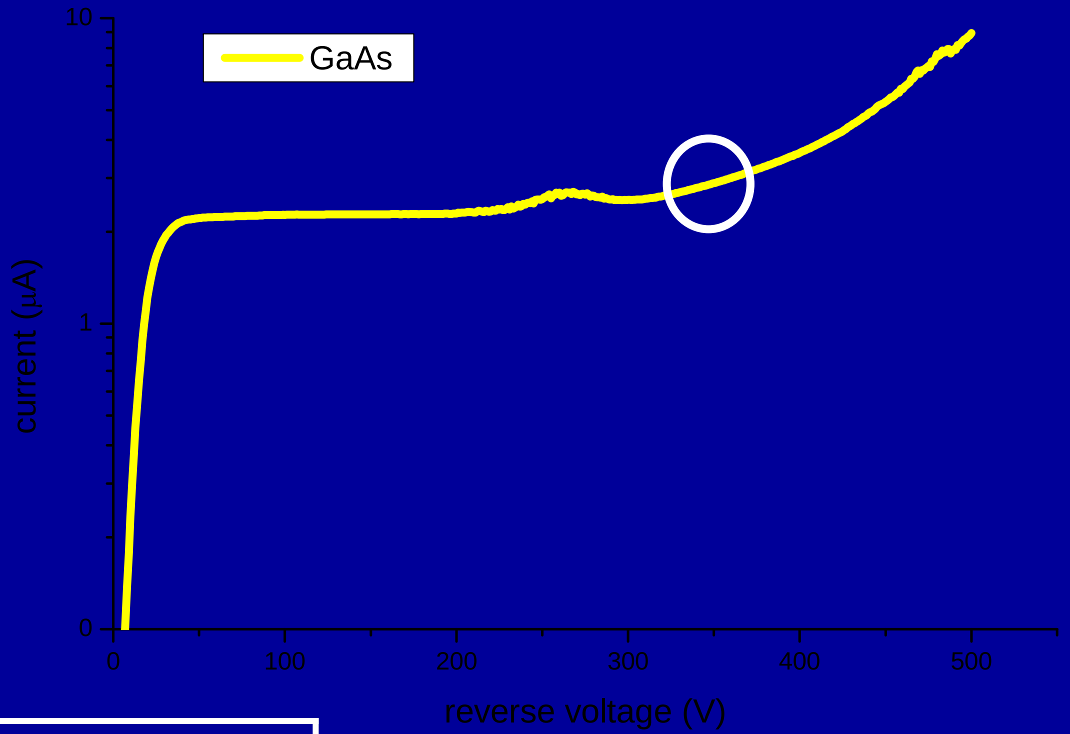



• The performance of the GaAs detector in terms of the SNR is higher than the 525 mm thick Si one

• In agreement with the different detection efficiency of Si and GaAs at these energies.

# Electrical characterization

The voltage has been applied on the back side up to the maximum value of 500 V.
(Tested for several days at room temperature)

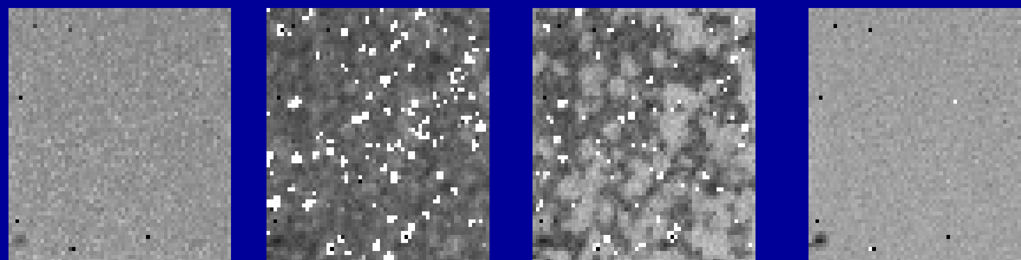



| | |
|---|---|
| **350V** | **Operating bias voltage** |
| **2.4 $\mu A/cm^2$** | **Current density** |

Current density is in agreement with the requirement (lower than 5 $\mu A/cm^2$)

# Working point optimization

Irradiation with a $^{109}Cd$ source



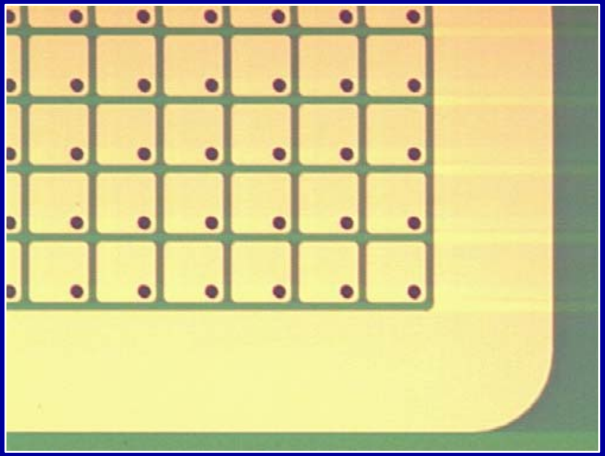
**100 V** **250 V** **300 V** **350 V**

100 V: uniform image but low detection efficiency
250 V and 300 V: inhomogeneity and degradation in the image quality
350 V: uniform image with high statistics and high efficiency

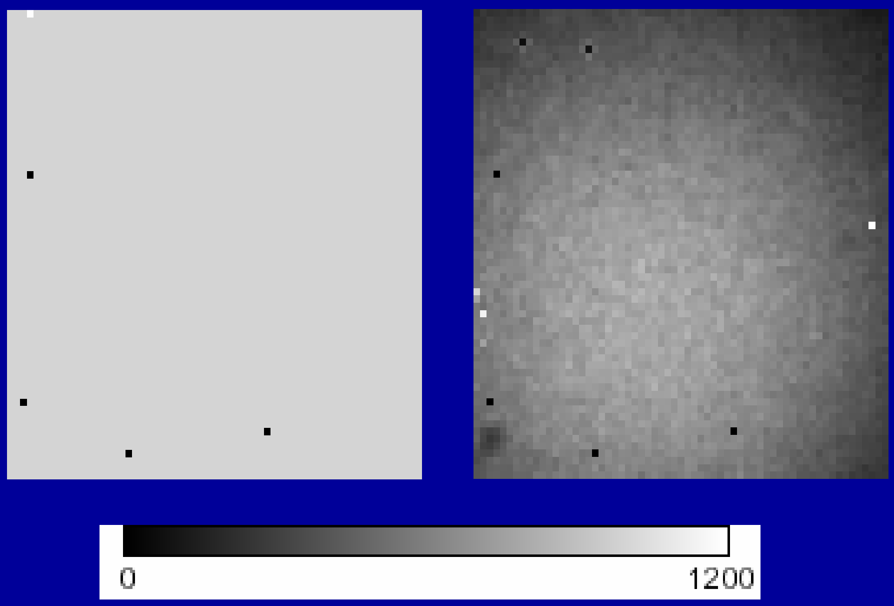
# The Indium Bump Bonding

- Goal
  - High quality electrical contact between GaAs detector and Si electronics
  - High bonding yield

- The AMS (Alenia Marconi Systems) process
  - Under Bump Metal (UBM): evaporation of a Ti/Al multilayer.
  - Bumps: Indium evaporation on the electronics and the detector pads
  - Bumps: cylindrical 30μm in diameter, 9μm high



# The bump-bonding tests




- For the electronics tests, black pixels are damaged
- For the tests with the Ru106 source, search for additional not working pixels (damaged or unconnected after the bump bonding)

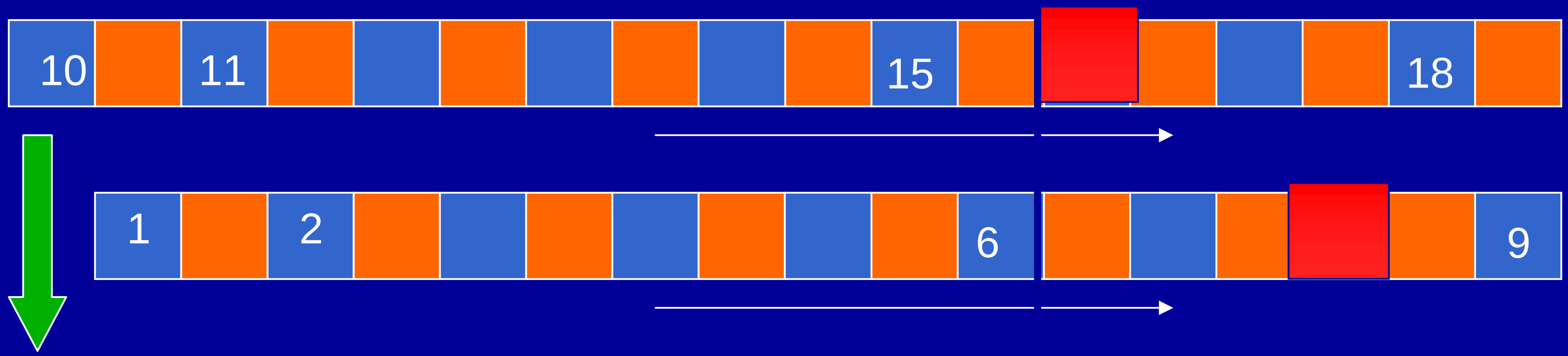
# The assemblies selection

- More than 40 GaAs assemblies have been produced and tested
- The best 18 GaAs assemblies have been integrated in the mammographic head.
- The number of bad pixels is between 30 and 180 for an overall yield between 95% - 99%.

| assembly | 1 | 2 | 3 | 4 | 5 | 6 | 7 | 8 | 9 | 10 | 11 | 12 | 13 | 14 | 15 | 16 | 17 | 18 |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| Nr. bad pixels | 30 | 30 | 40 | 50 | 50 | 50 | 60 | 65 | 70 | 70 | 100 | 110 | 130 | 130 | 150 | 160 | 160 | 180 |
| Overall yield | 99.3 | 99.3 | 99.0 | 98.8 | 98.8 | 98.8 | 98.5 | 98.4 | 98.3 | 98.3 | 97.6 | 97.3 | 96.8 | 96.8 | 96.3 | 96.1 | 96.1 | 95.6 |

# GaAs Detectors Test

- The 18 assemblies mounted onto the chipboard have been biased at 350 V and the inverse current drawn by each one have been measured
- 16 chips over 18 draw a current less than 5 μA, which is the limit fixed to prevent the detector breakdown




- we can use only a region 12 cm wide of contiguous good chips. The chips 8 and 16 are unusable due to the high leakage current.
- usable assemblies: from No. 1 to No. 6 (bottom row) and from No. 10 to No. 15 (top row)

# Set-up optimization

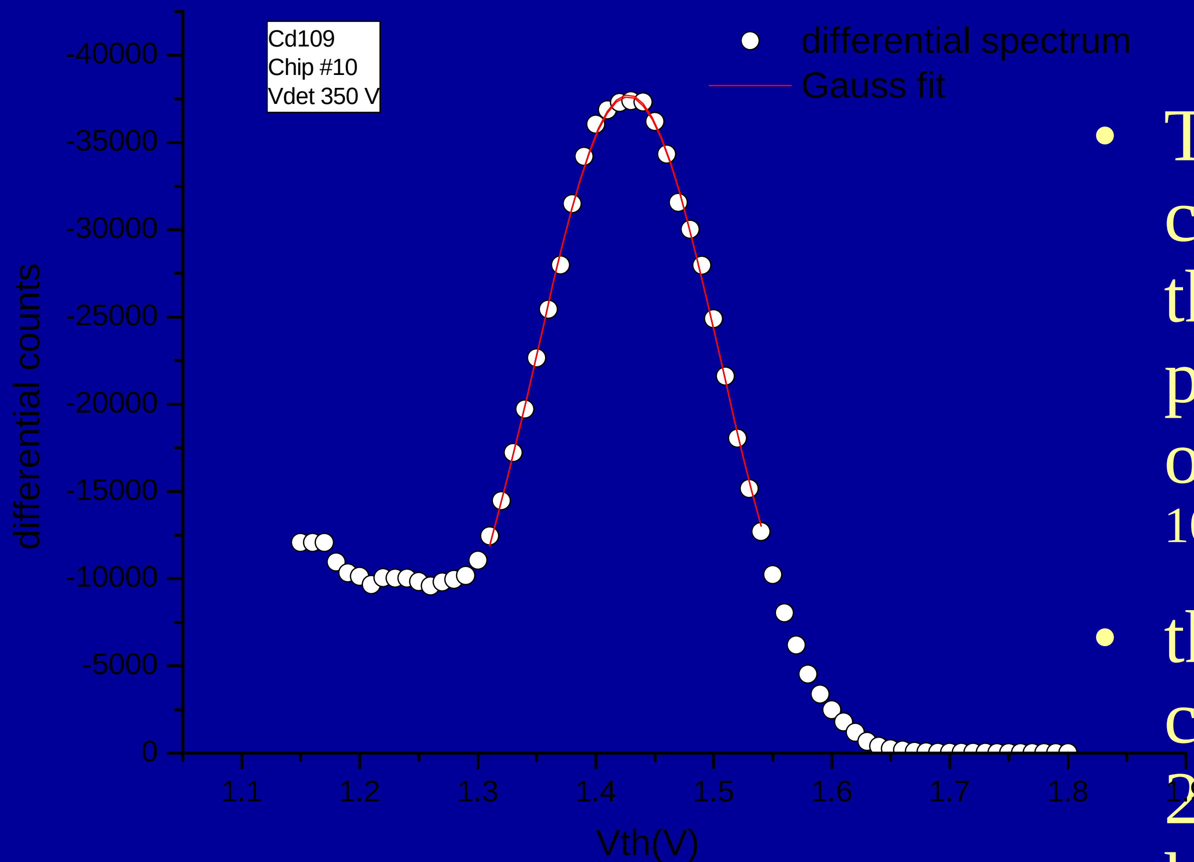
- MedipixI/PCC's electrical characterization
  - Working point optimization
  - Threshold adjustment of the single chips
  - Absolute and electrical calibration of the 12 "good" assemblies thresholds
  - Threshold equalization among the 12 assemblies
- Beam tests with the mammographic tube
  - Thresholds-counts correlation
  - Radiographs of a mammographic phantom with low contrast details

# Electrical Calibration

- The equivalent threshold of each chip (average amplitude in mV) has been determined by electrical pulses sent through the test input of the MPXI/PCCs as a function of the threshold value Vth

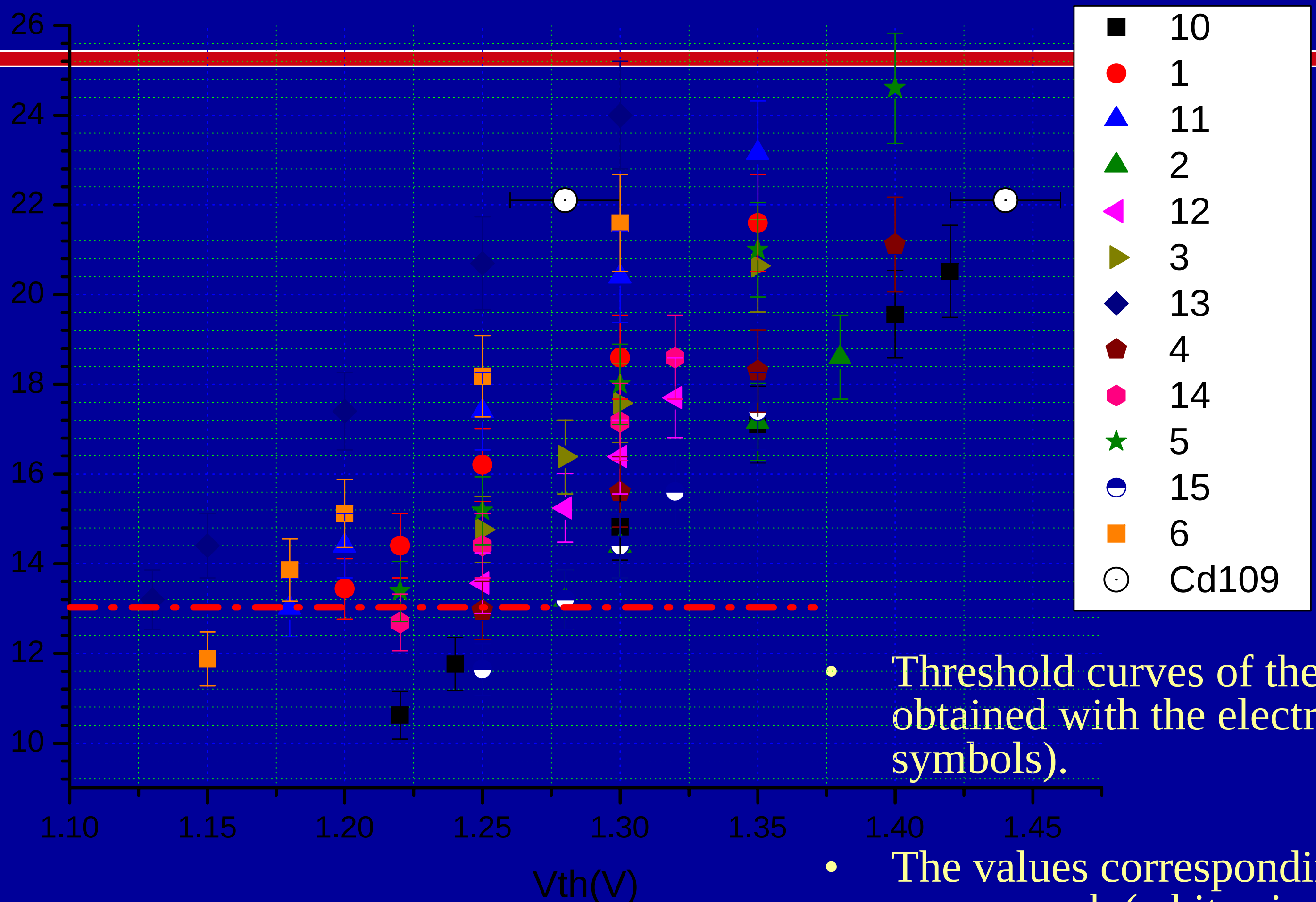
- This procedure has been repeated for different threshold values Vth and for all the 12 good assemblies

# Absolute calibration




- The absolute calibration of the thresholds in terms of photon energy has been obtained by using a $^{109}Cd$ gamma source
- the threshold corresponding to the 22 keV emission peak has been determined by acquiring a discrete differential spectrum.

# Calibration Curves




- Threshold curves of the 12 assemblies obtained with the electrical pulses (colored symbols).
- The values corresponding to the $^{109}Cd$ 22 keV energy peak (white circles) for two different chips (assemblies 10 and 13) are altogether reported.
- The horizontal dot-dashed line represents the average settable minimum threshold.

# Threshold equalization among the 12 chips

- For every chip, we can fix a minimum threshold below which the chip becomes noisy. The minimum threshold value depends on the number of noisy pixels that can be accepted.

- In our case, the assembly reaches the minimum threshold when in a dark acquisition (no X ray exposure) of 1 second the maximum number of noisy pixels is 10.

- Upon this criterion, the minimum threshold of the 12 chips ranges from 11 keV (chip 10) to 16 keV (chip 3).

- If we exclude the chip with the highest threshold (chip 3), the average threshold that can be set for all the chips is about13 keV

# Mammographic Phantom

Lucite cylinder

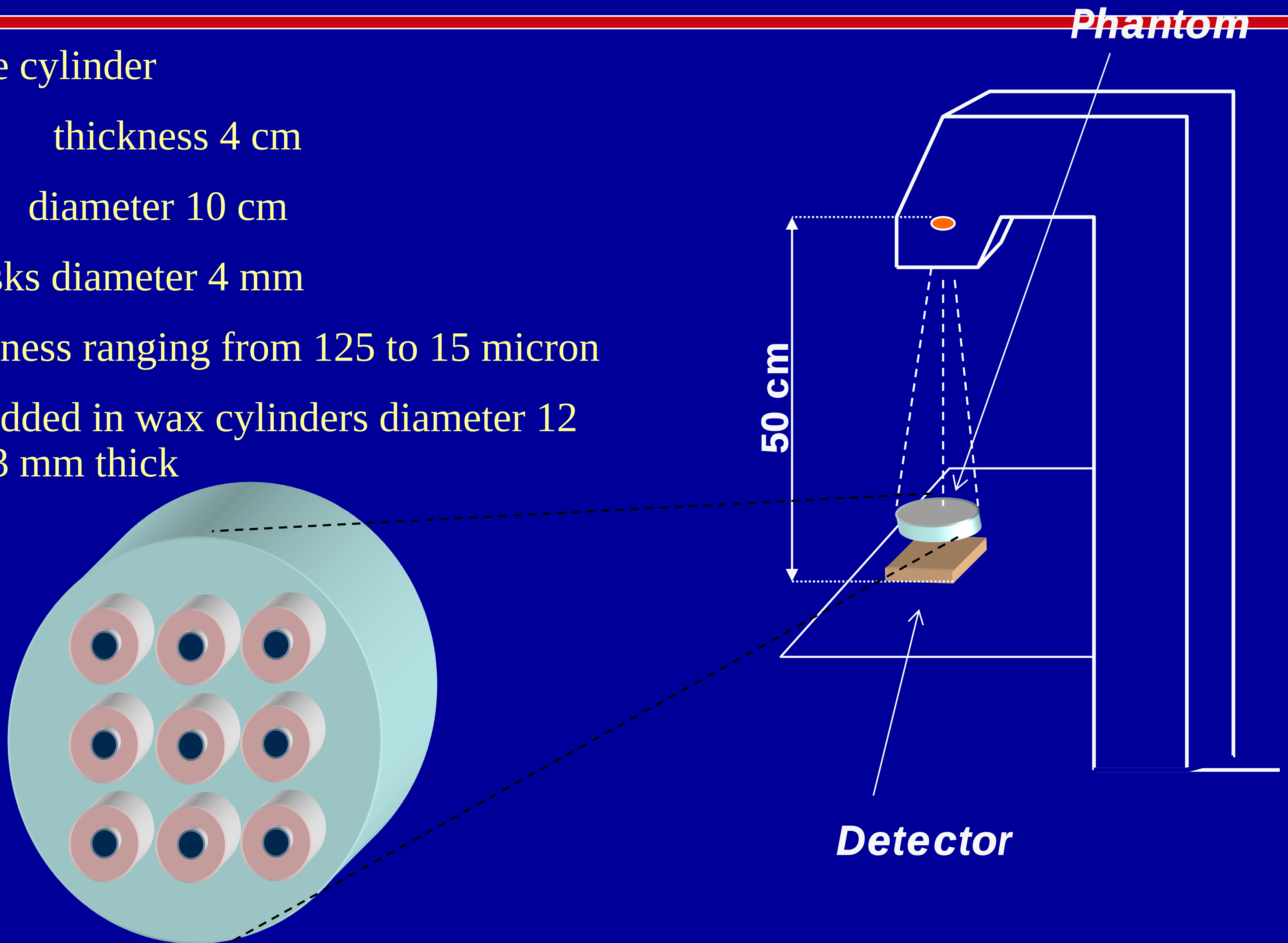
thickness 4 cm

diameter 10 cm

Al disks diameter 4 mm

Thickness ranging from 125 to 15 micron

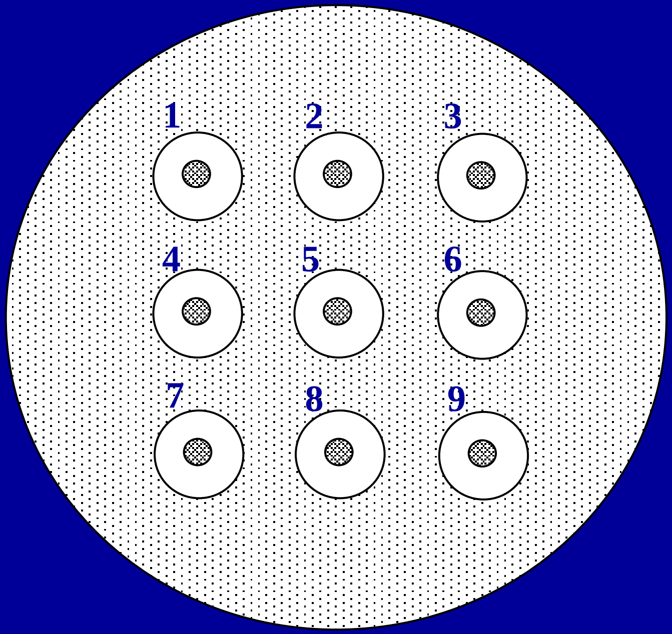
Embedded in wax cylinders diameter 12 mm, 3 mm thick



# Phantom Radiograph

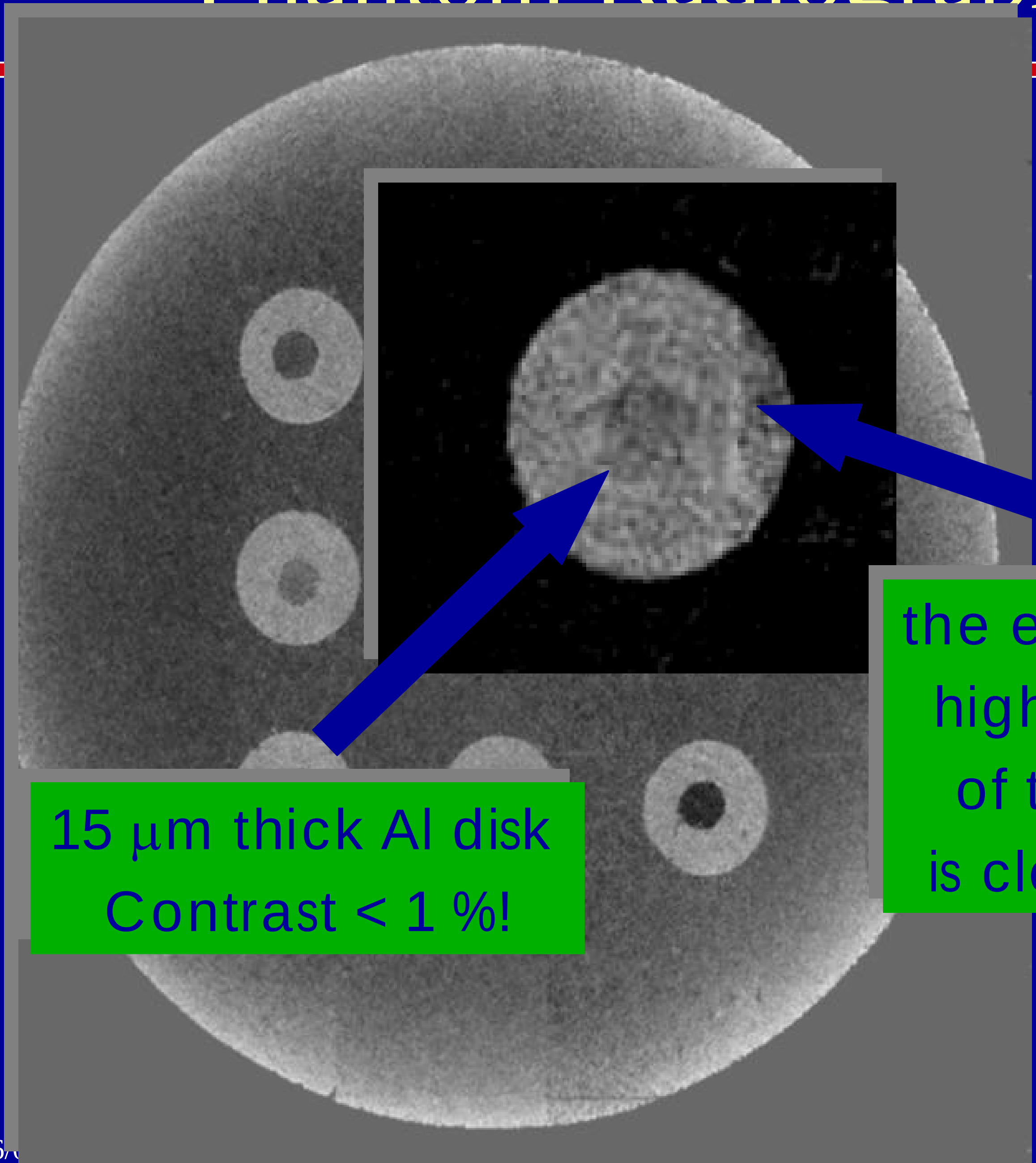
- The phantom was exposed to the X ray beam and the image acquired by scanning the detector across the phantom surface with a sequence of 13 exposure-motion steps.
- Mammographic Tube Settings:
  - 27 kV, 30 mAs, focus 300 μm, exposure field 24 x 30 $cm^2$
- Acquisition times
  - Shutter time: 0.5 s
  - Delay time: 20 s
  - Limited by heat dissipation rate from the mammographic tube
- The dose per exposure (measured with a silicon dosimeter) delivered to the phantom is 4.8 mGy.
- The image of the phantom has been obtained by weighting the raw image with a high statistics flat field and corrected for the noisy pixels.

# The Phantom




| Part # | Thickness (μm) | Contrast | SNR |
|---|---|---|---|
| 1 | 75 | 0.040 | 3.85 |
| 2 | 15 | 0.009 | 1.03 |
| 3 | 50 | 0.028 | 2.74 |
| 4 | 50 | 0.027 | 2.57 |
| 5 | 40 | 0.022 | 2.19 |
| 6 | 100 | 0.054 | 4.48 |
| 7 | 25 | 0.015 | 1.63 |
| 8 | 100 | 0.048 | 4.86 |
| 9 | 125 | 0.070 | 5.77 |

# Phantom Radiograph



15 μm thick Al disk
Contrast < 1 %!

the effect of the high threshold of the chip 3 is clearly visible

# Bar Patterns

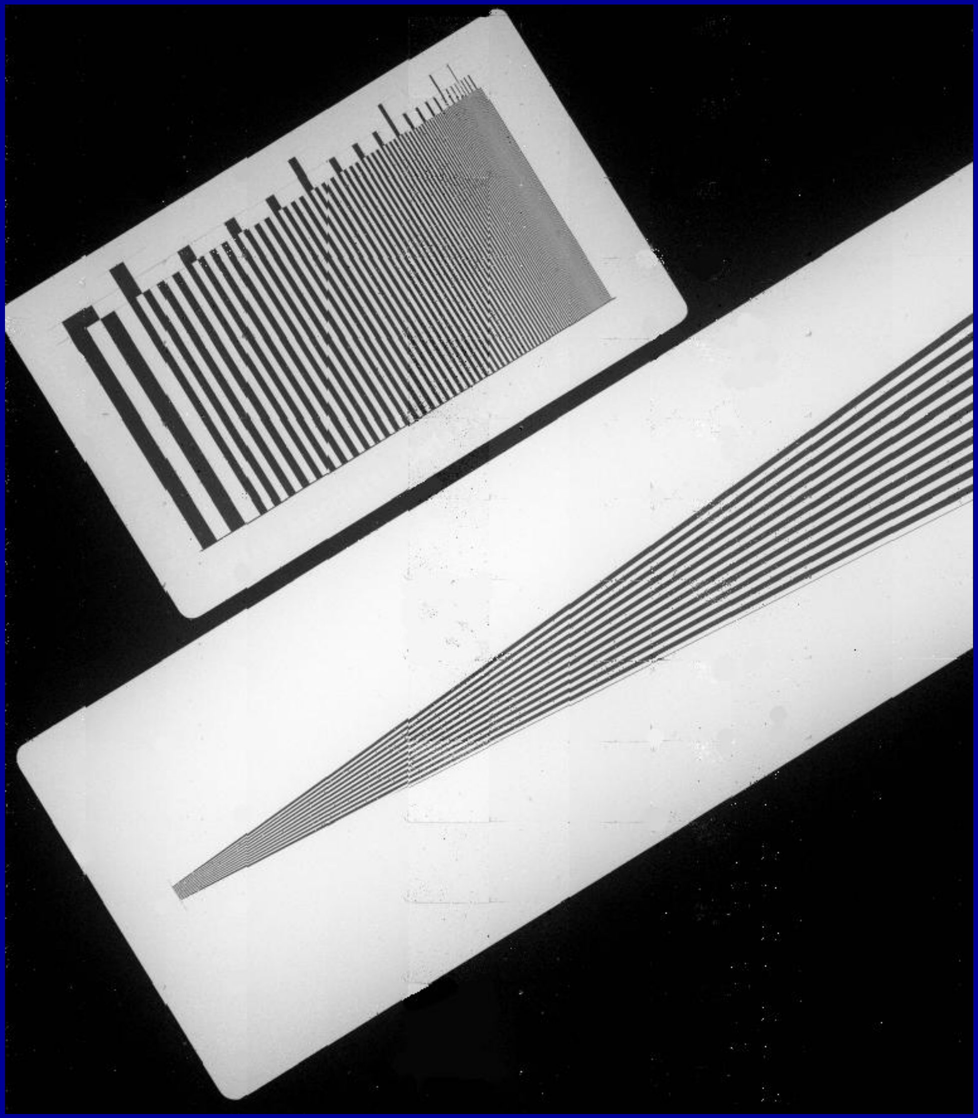


Image of two bar patterns (from 0.5 to 10 lp/mm) acquired to verify the alignments of the chips and the accuracy of the stepper motor

# A very low contrast image

The observable slight discontinuity in the image is due to the time stability of the X-ray tube output and of the detector thresholds, both of the order of 0.5%

# Conclusions

- The demonstrator has been fully tested and optimized to acquire radiographs of a mammographic phantom in exposure conditions typical of a clinical examination.
- The system is able to produce good quality radiographic images with a standard dose.
- Its performance is particularly high in the detection of low contrast details.
  - With this system, we were able to detect a 15 μm thick Al disk against a wax background. The contrast measured on the image is $< 1$ %.

# Acknowledgments

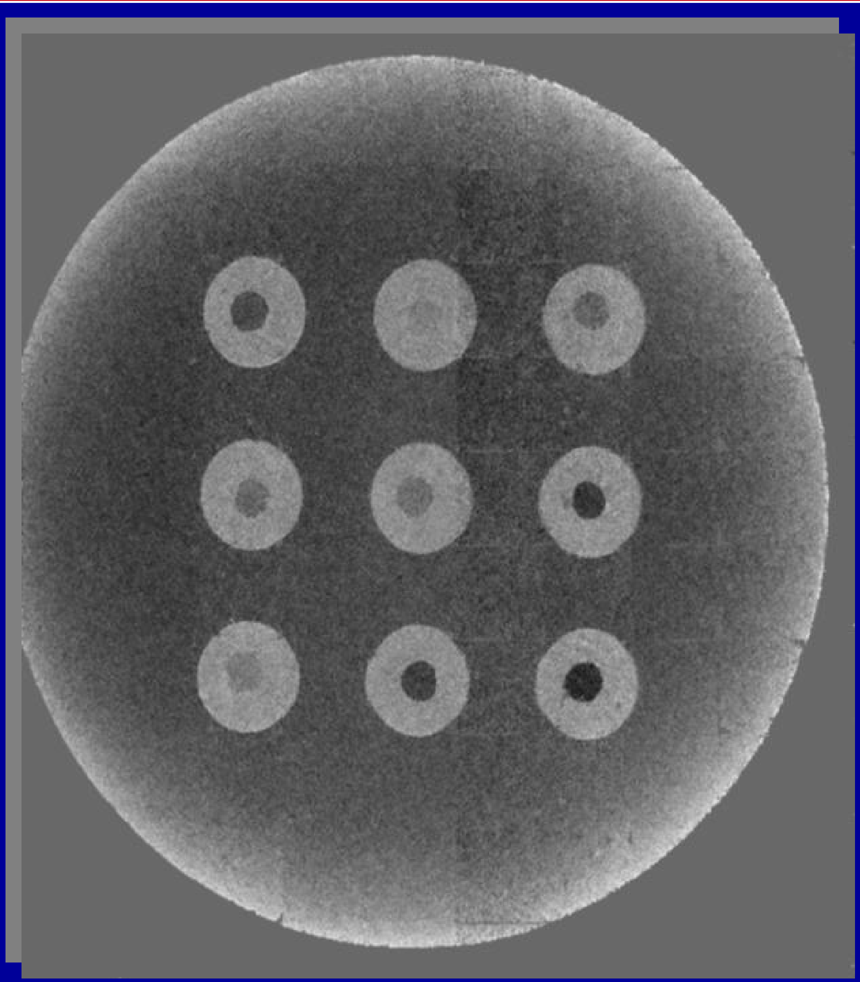
This project has been supported by the

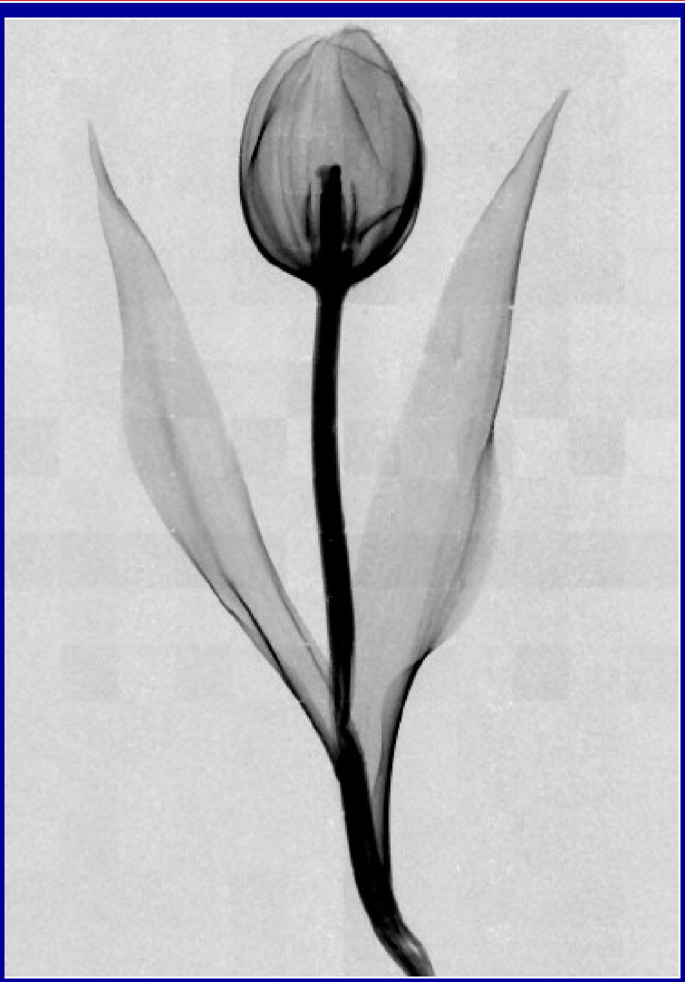
under the Law 46/art. 10-1996

and by the Italian High Tech companies

in collaboration with

# Image of a very low contrast object





- The dose per exposure (measured with a silicon dosimeter) delivered to the phantom is 4.8 mGy.
- The image of the phantom has been obtained by weighting the raw image with a high statistics flat field and corrected for the noisy pixels.

Università di Pisa
**Facoltà di Scienze Matematiche Fisiche e Naturali**
**Corso di Laurea in Fisica**
Anno Accademico 2005/2006

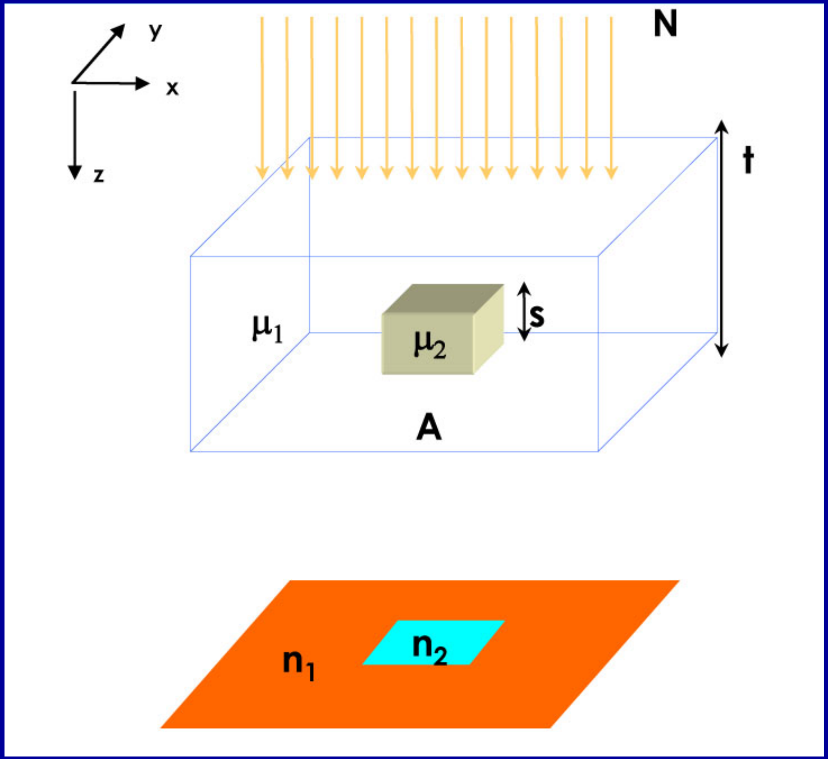
Elaborato Finale

# Caratterizzazione di un sistema per mammografia digitale, basato su chip a conteggio di singolo fotone e rivelatori a GaAs

Candidato
Giovanni Paternoster

Relatore
Dott. M. G. Bisogni

# L’indagine mammografica




Energie tipiche dell’indagine mammografica: 10 – 30 keV

Contrasto (C) e Rapporto Segnale Rumore (snr), sono i 2 principali parametri qualitativi di un’immagine mammografica

$$C = \frac{n_1 - n_2}{n_1} = 1 - e^{-(\Delta\mu)s}$$

$$snr = \frac{n_1 - n_2}{\sqrt{\sigma_1^2 + \sigma_2^2}}$$

# Il Dimostratore

Tubo mammografico

Collimatore

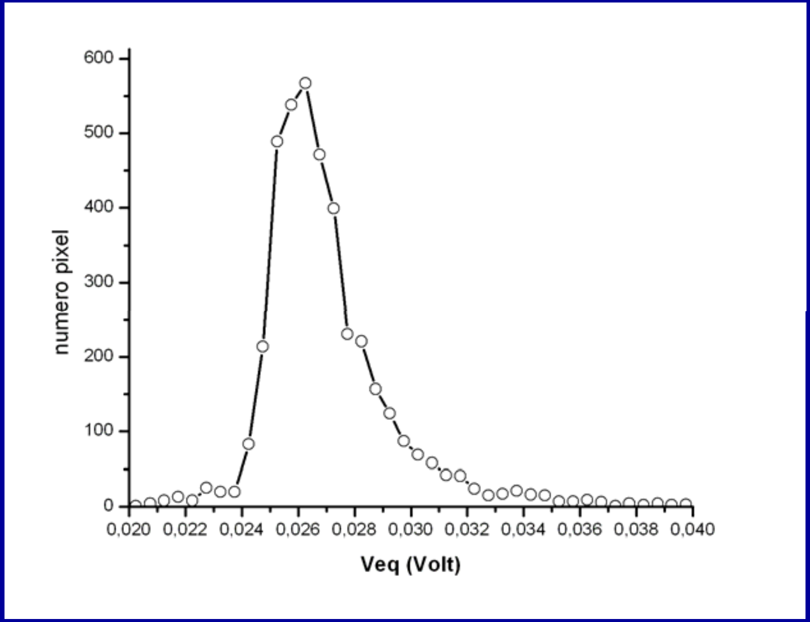
Sistema di raffreddamento

Sistema di rivelazione

Connettori di alimentazione
e connessione al PC

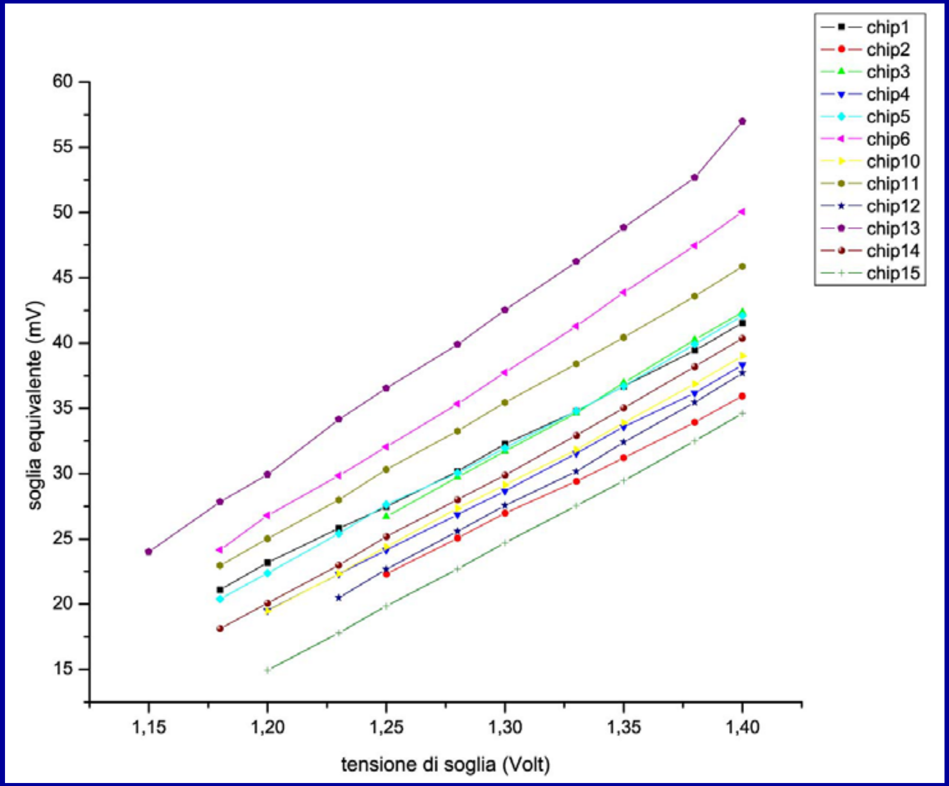
# Calibrazione delle unità di rivelazione

- **Calibrazione elettrica ed assoluta delle soglie dei 12 chip contigui.**
- **Equalizzazione delle soglie dei 12 chip contigui.**



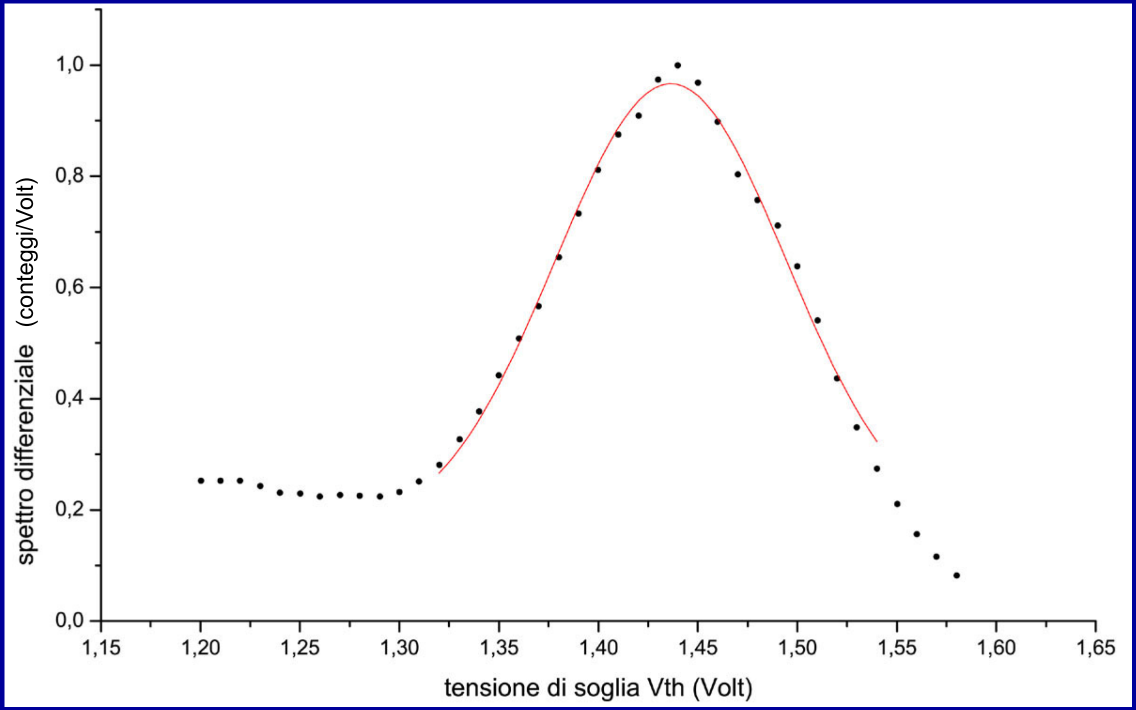

**In figura mostriamo la distribuzione delle soglie equivalenti dei pixel del chip 1.**

# Calibrazione delle soglia




- **Tramite impulsi elettrici di tensione inviati al test input di un pixel del chip Medipix, è possibile determinare la soglia equivalente di ciascun pixel (ampiezza media in mV) in funzione della tensione di soglia Vth. E' quindi possibile stimare la media della soglia equivalente sui pixel di un chip in funzione di Vth.**

# Calibrazione assoluta




- **Tramite una scansione della tensione Vth, viene ricostruito lo spettro di emissione di una sorgente gamma di $^{109}Cd$ (E = 22 keV)**
- **In figura lo spettro sperimentale acquisito dall'unita di rivelazione No. 12**

# Equalizzazione delle soglie

**Data la linearità tra la soglia equivalente e il Vth e tra la soglia equivalente e la soglia energetica, è possibile determinare la forma di E(Vth) ed esprimere le soglie di discriminazione in energia (keV)**

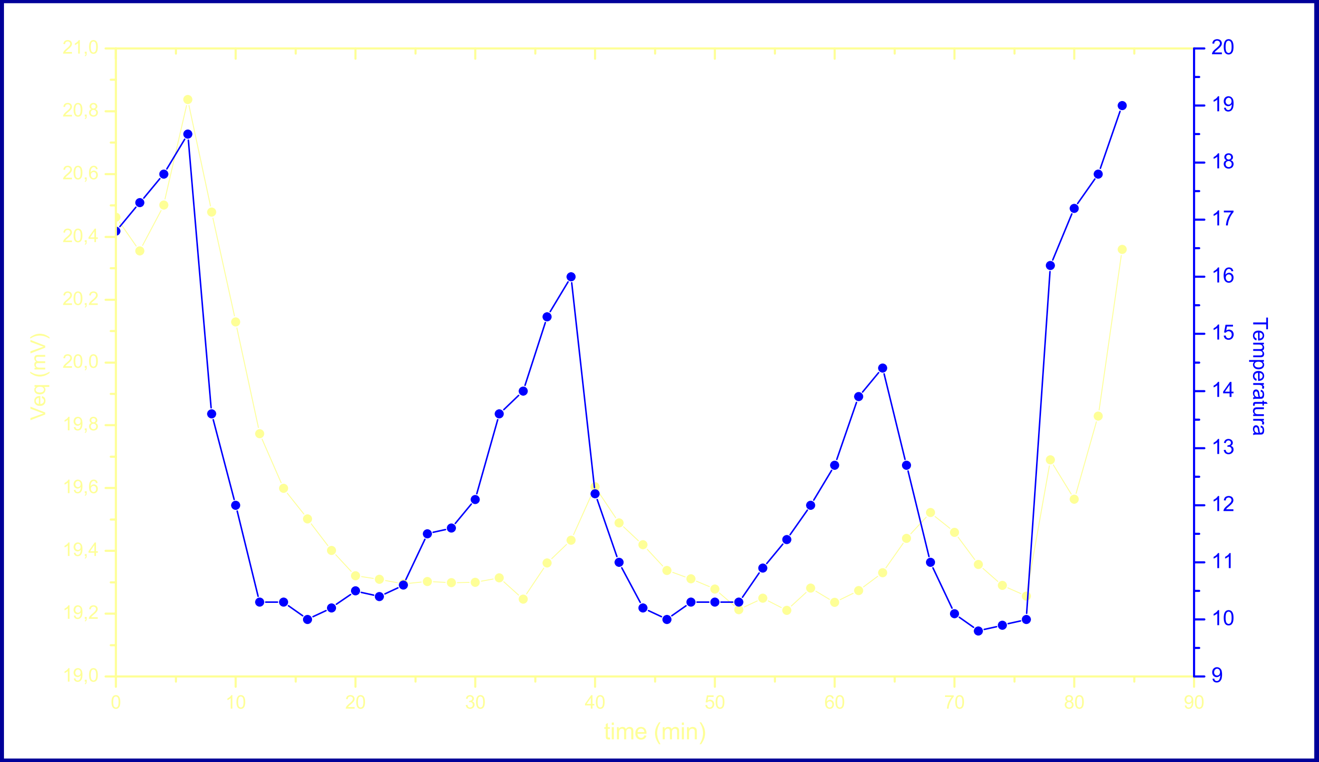
**Impostiamo le soglie energetiche di 11 chip al valore di 13 keV.**
**La soglia energetica del chip 3 invece è impostata a 15 keV a causa dell’alto numero di pixel rumorosi.**

# Stabilità del sistema

**Per acquisire un’immagine a campo pieno il sistema effettua 26 scansioni consecutive, intervallate da 10 secondi così da permetter il raffreddamento del tubo RX.**

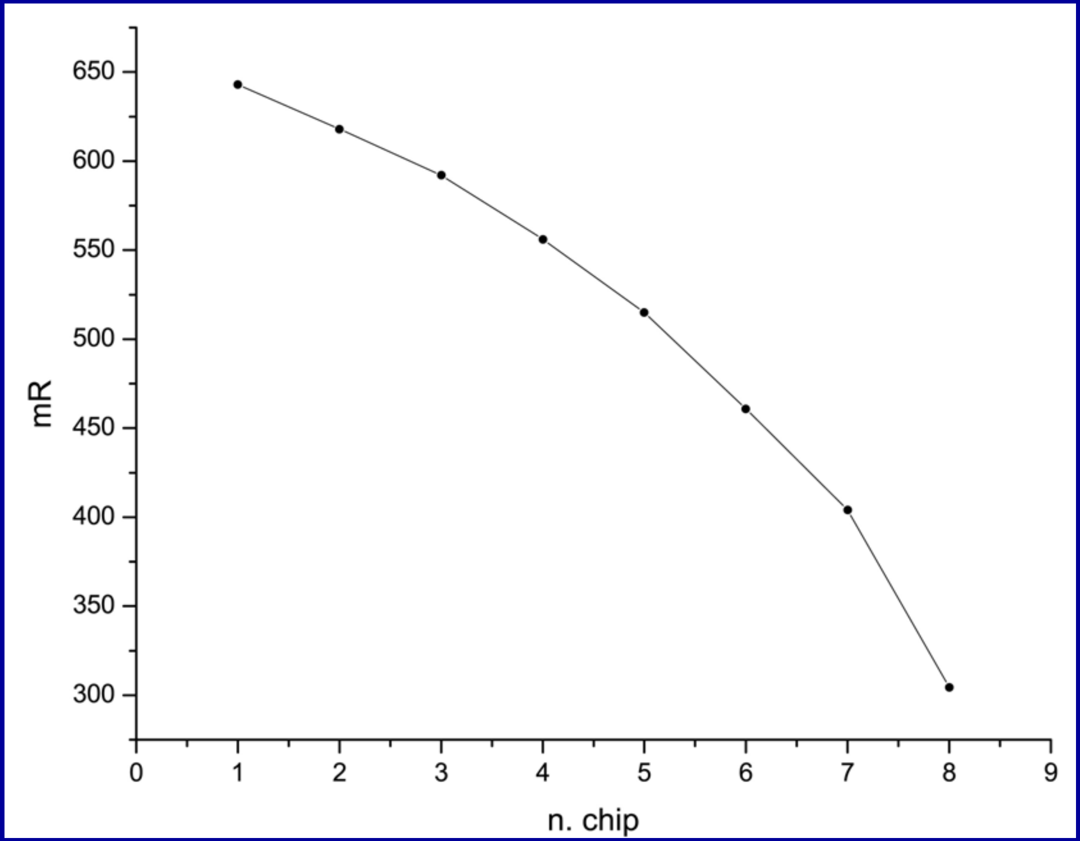
**Ci interessa verificare che in questo intervallo di tempo la risposta del sistema non vari.**

# Costanza delle soglie




**La soglia equivalente ha un andamento in accordo con quello della temperatura.**
**Una variazione delle soglie dello 0.2 % comporta una variazione nei conteggi dello 0.5 %**

# Caratterizzazione del tubo RX




**Il fascio emesso dal tubo RX non è uniforme su tutto il piano di rivelazione. Misure effettuate con camera a ionizzazione mostrano l'andamento in figura.**

**Inoltre la dose in aria misurata in diverse esposizioni consecutive non è costante.**

**Questo comporta una differenza nei conteggi tra due acquisizioni consecutive.**

# Caratterizzazione del tubo RX

**Misure della dose in aria effettuate con camera a ionizzazione mostrano che l'output del tubo non è costante.**
**Misure su 40 esposizioni consecutive danno:**

**Media 818.0 mR   std: 1.9mR**

**Variazione massima tra 2 esposizioni consecutive : 4 mR**

**Variazione della dose in aria dello 0.5 %** → **Analoga variazione nei conteggi di due acquisizioni**

# Stabilità del sistema

**In definitiva la variazione delle soglie e la non costanza dell'output del tubo determinano una variazione nei conteggi tra due scansioni consecutive fino all'1 %.**

# Equalizzazione delle immagini

**Per minimizzare l'errore sistematico dovuto alla disuniformità nella risposta dei pixel, e del rivelatore, nonché alla disuniformità del fascio RX sul piano di rivelazione, si equalizza l'immagine con un fondo, ovvero un'immagine acquisita a campo piatto.**

$[I]_{i,j}$ Conteggi registrati da ciascun pixel

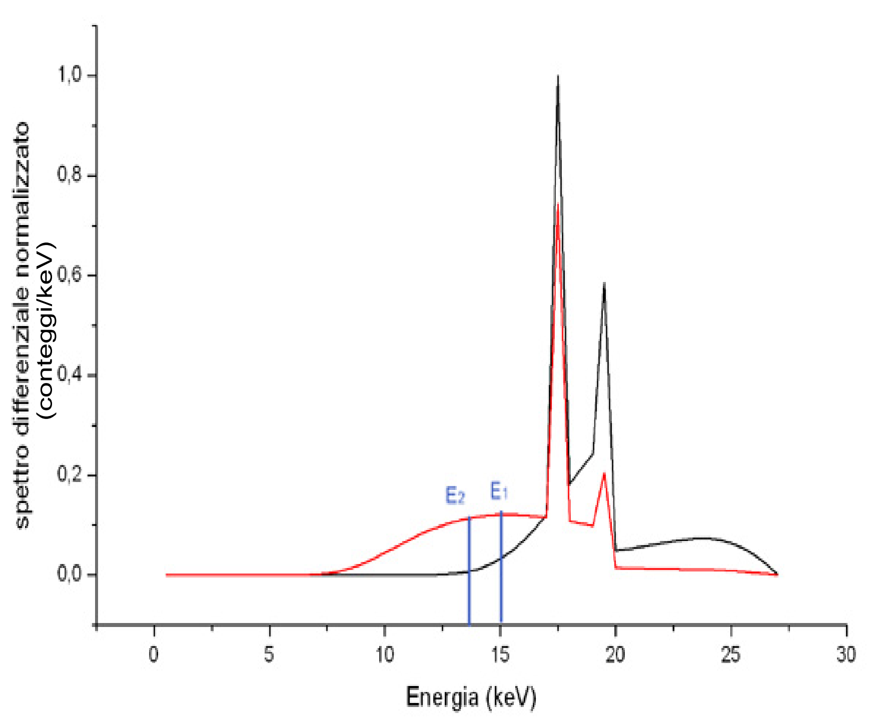
**IMMAGINE [I]**

**FONDO [F]**

**IMMAGINE EQUALIZZATA [C]**

$$[C]_{ij} = \frac{[I]_{ij}}{[F]_{ij}} < [F]_{ij} >$$

# Equalizzazione delle immagini

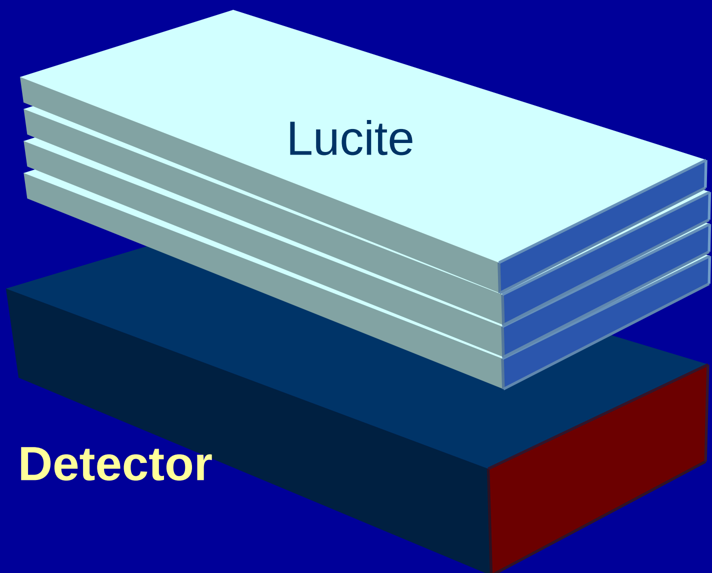


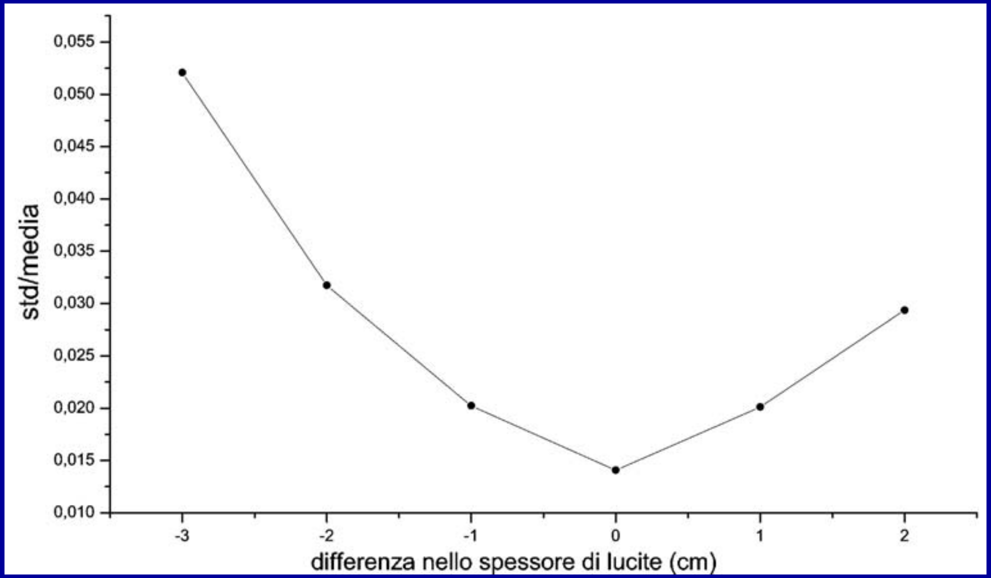

Il fondo di equalizzazione è strettamente dipendente dallo spettro usato per acquisirlo. Se gli spettri normalizzati incidenti sui rivelatori al momento dell'acquisizione del fondo e dell'immagine sono diversi, nel processo di equalizzazione si introduce un errore sistematico.

**I due spettri energetici in figura rappresentano la radiazione incidente sui rivelatori all'acquisizione dell'immagine e del fondo**

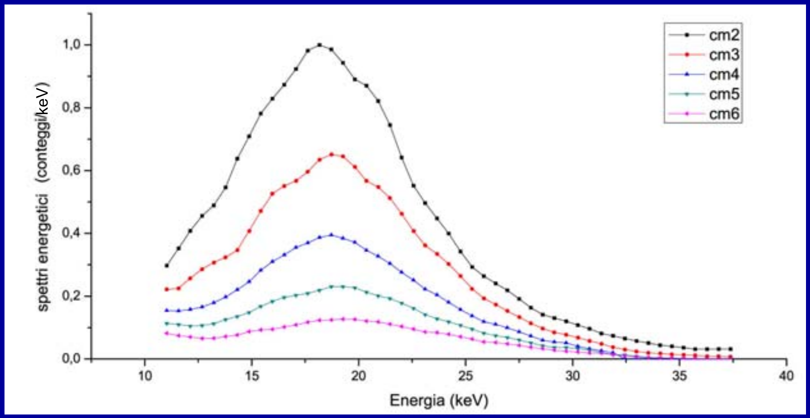
# Equalizzazione delle immagini

**Equalizziamo un'immagine acquisita ponendo 4 cm di lucite sul rivelatore con dei fondi acquisiti utilizzando diversi spessori di lucite**

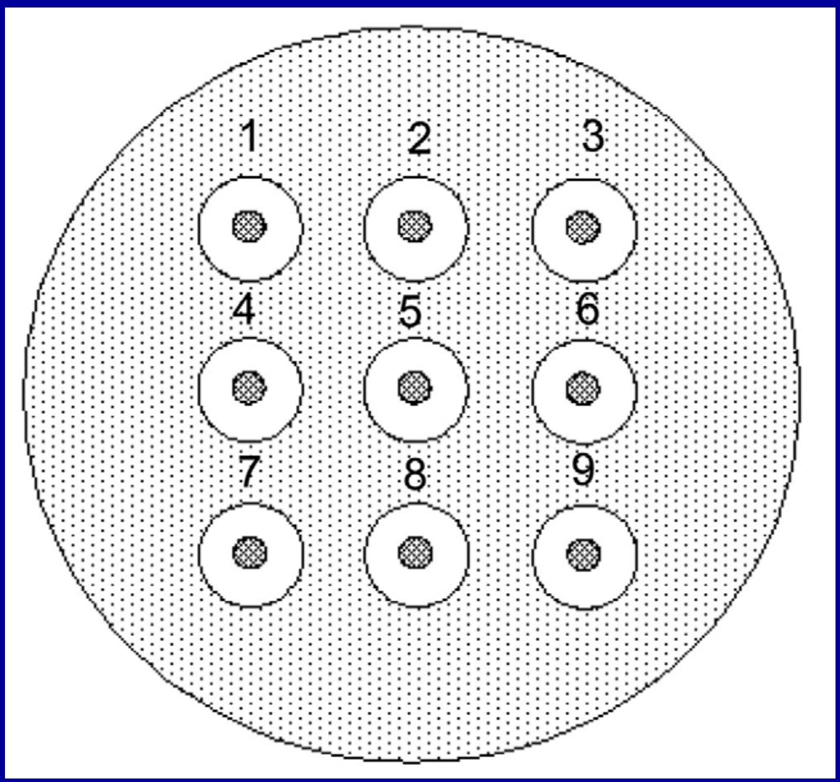



**In figura gli Spettri sperimentali del fascio emesso dal tubo RX rivelati dall'unità 12 a diversi spessori di lucite**



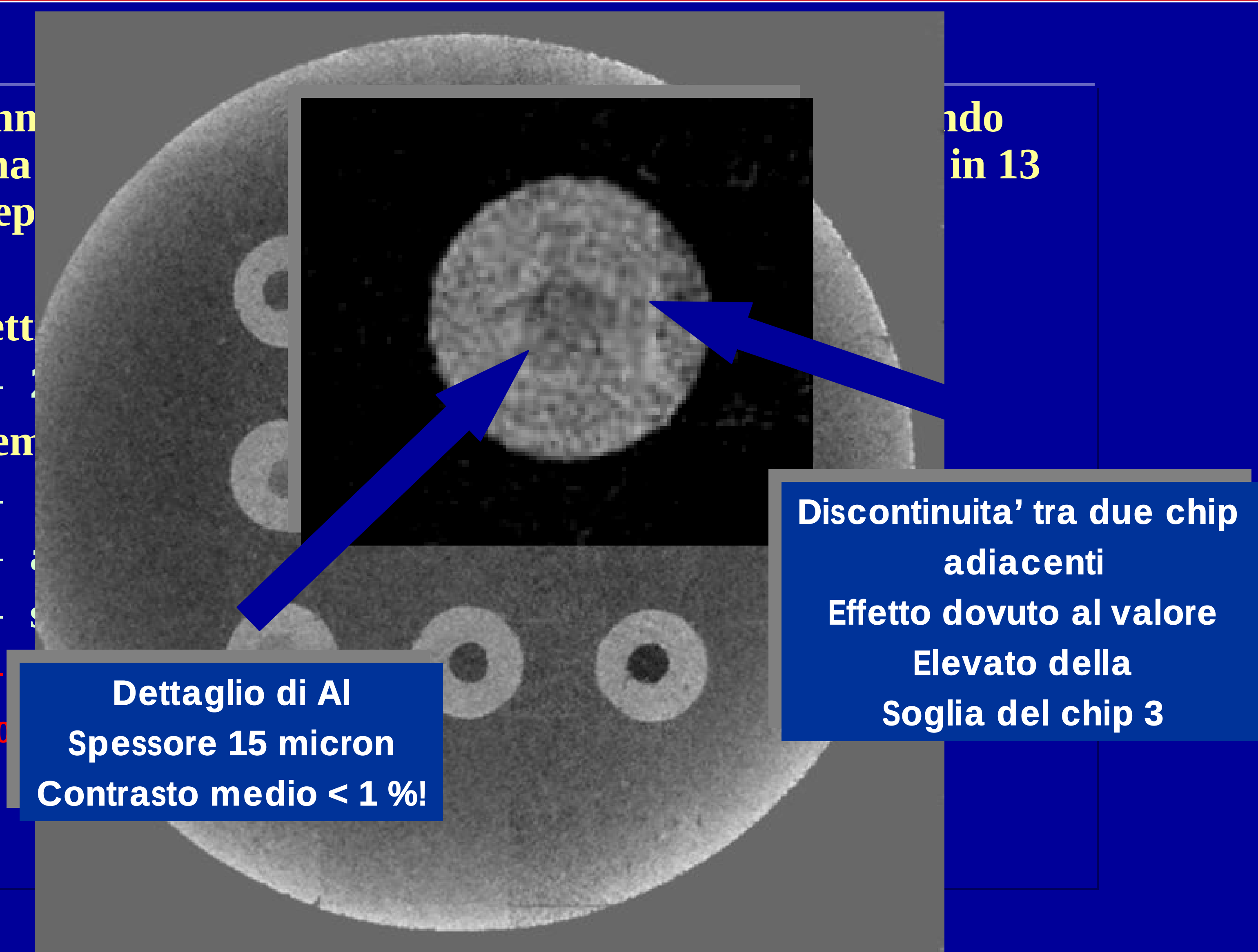
# Fantoccio mammografico




**Il fantoccio è costituito da un cilindro di raggio 5 cm e spessore 4 cm in lucite, nel quale sono stati inseriti 9 cilindri di cera, di circa 1 cm di diametro e contenenti dischetti di Al di circa 4 mm di diametro e di diversi spessori (da 125 micron a 15 micron).**

# Immagine del fantoccio

- Imm… ndo una… in 13 step…
- Sett…
  - 2…
- Tem…
  - …
  - a…
  - s…
  - …
- Do…



# Conclusioni

- Il sistema è stato ottimizzato per acquisire immagini mammografiche di fantocci in condizioni di esposizione tipiche di una mammografia clinica.
- Il sistema è in grado di produrre immagini mammografiche di ‘buona qualita’.
- Le sue prestazioni sono particolarmente elevate nella rivelazione di dettagli a basso contrasto

# Conclusioni

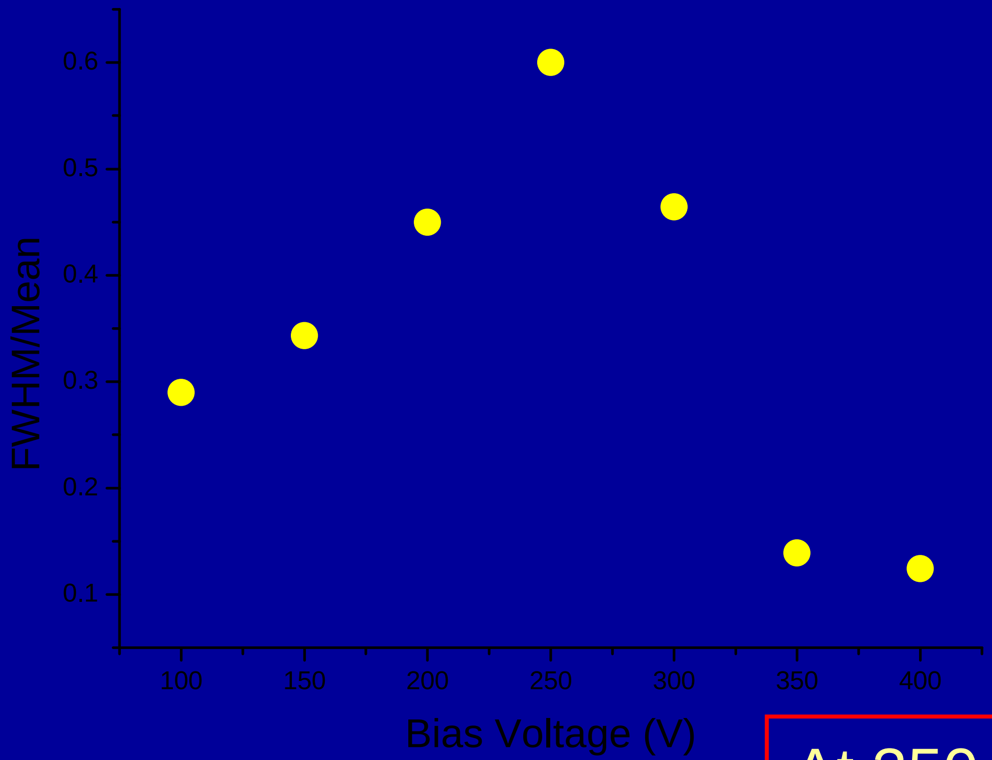
- Il chip n. 3 è rumoroso e va sostituito

**Tuttavia:**

- Il sistema necessita di un isolamento dall'esterno e di un adeguato impianto di raffreddamento così da limitare la variazione delle soglie.
- La non costanza dell'output del tubo può essere corretta monitorando le diverse scansioni con camera a ionizzazione e usando le misure per normalizzare l'immagine

# Working point optimization

Irradiation with X- ray tube (W anode at 40kV)

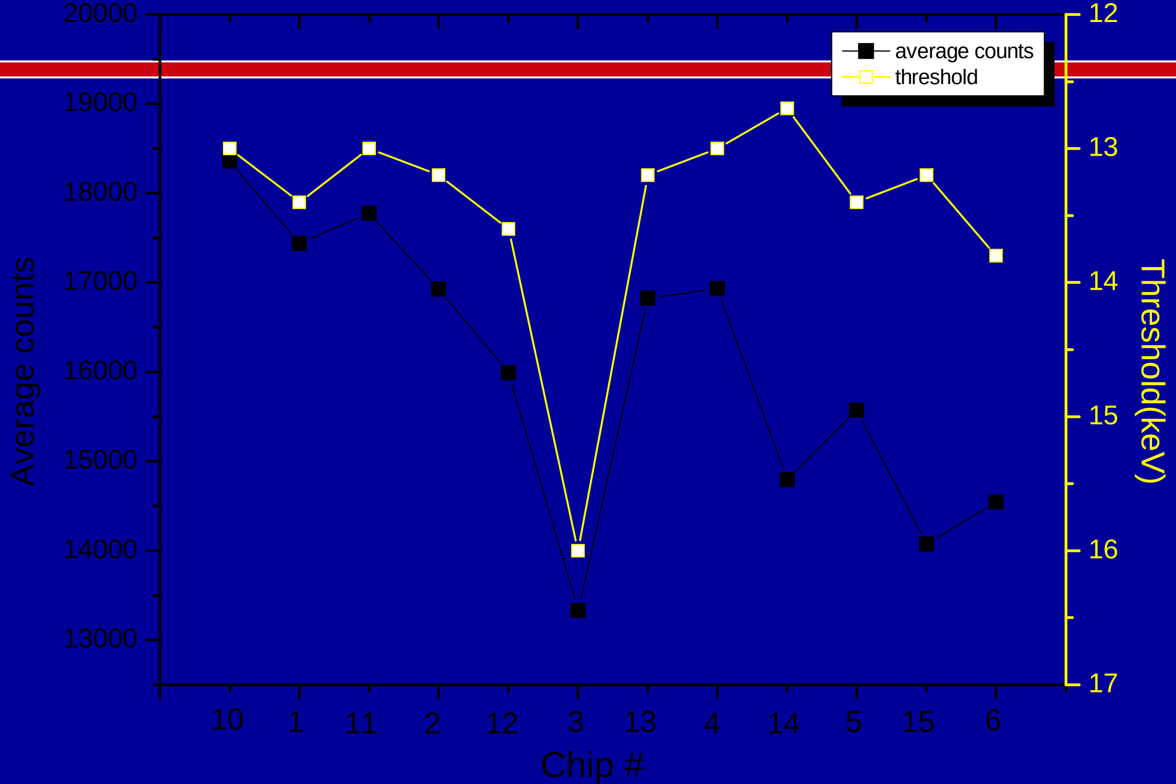



The ratio FWHM/mean provides a measure of the inhomogeneity degree among the pixels as function of the reverse voltage.

At 350 V:

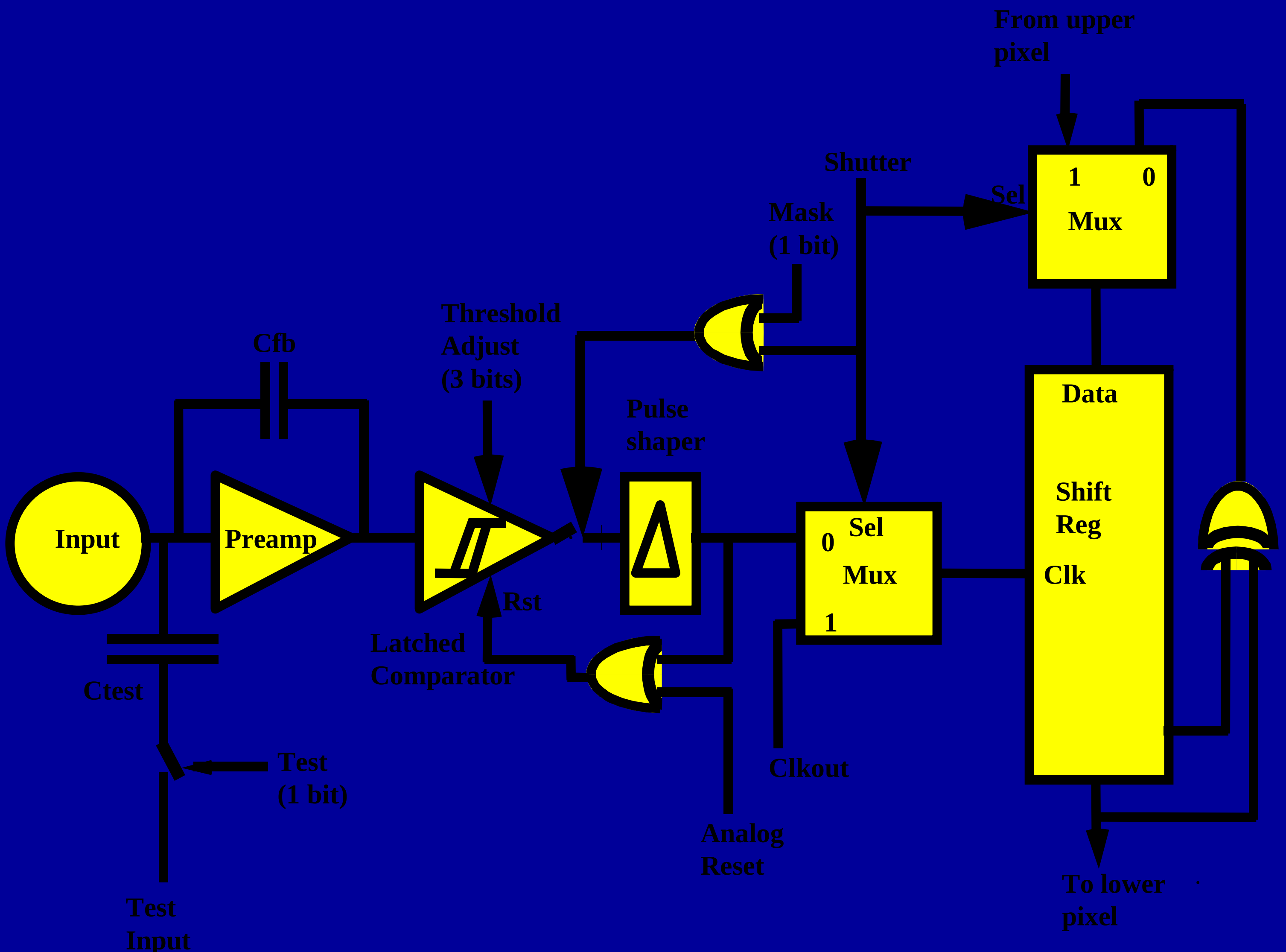
- The detector thickness is fully active
- The image is uniform
- The current density is 2.4

# Thresholds-counts correlation



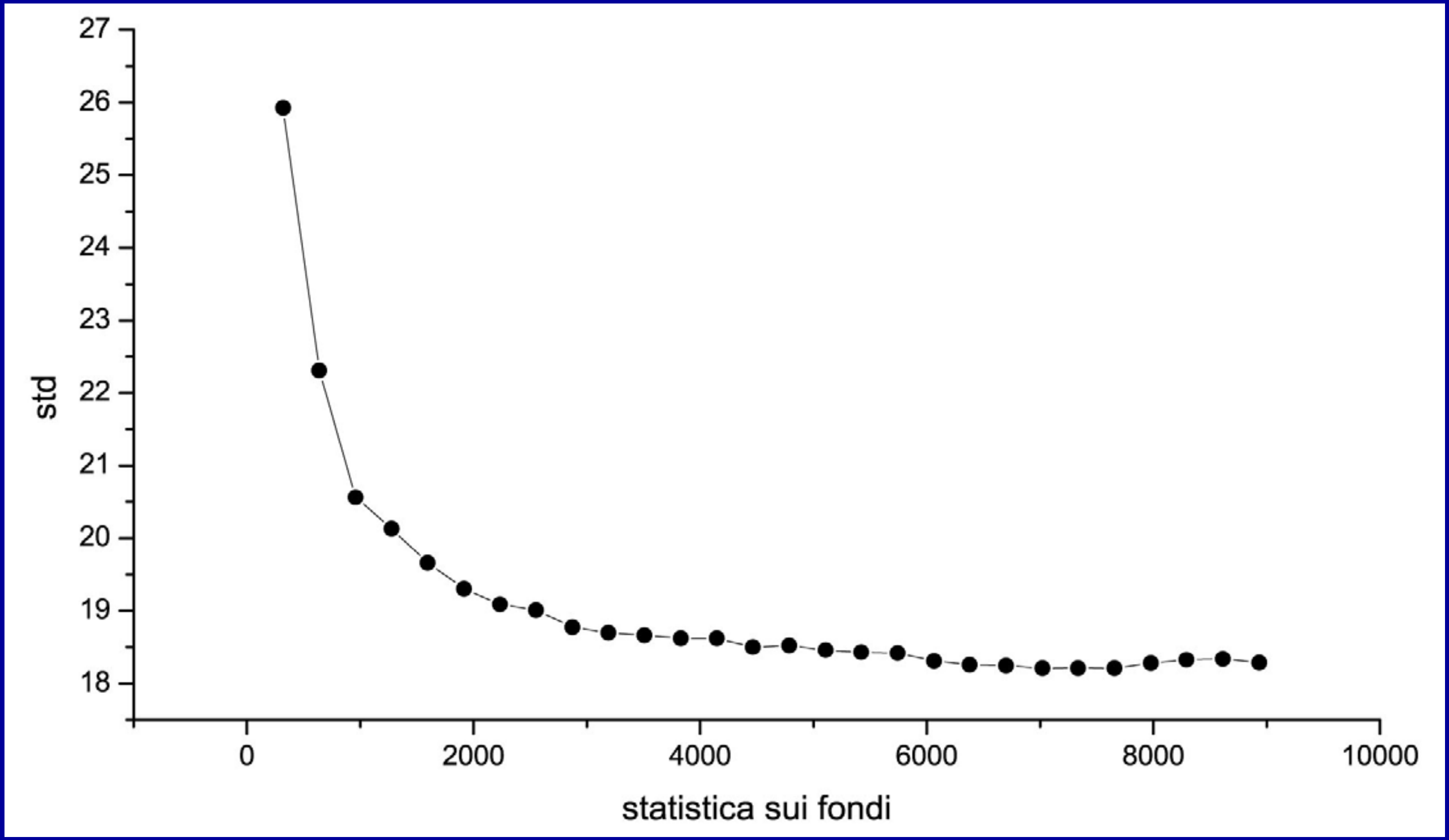

- Detectors exposed to the mammographic X-ray beam. The tube settings (27 kV, 30 mAs) same as in a clinical examination. 4 cm thick 18 x 24 $cm^2$ lucite slab onto the detectors to simulate breast tissue.
- the agreement between the two data sets is satisfactory, unless for the last four chips, where the counts are less than expected.
- This can be explained taking into account the “heel effect”. The intensity of the beam decreases across the exposure plane (from the detector 1 to the detector 6) due to the absorption in the anode heel.

# Medipix1



From upper pixel
Shutter
Sel
1
0
Mux
Mask (1 bit)
Threshold Adjust (3 bits)
Cfb
Pulse shaper
Data
Shift Reg
Clk
Input
Preamp
Sel
0
Mux
1
Rst
Latched Comparator
Ctest
Test (1 bit)
Clkout
Analog Reset
To lower pixel
Test Input

# Acquisizione dei fondi




**Std dell'immagine corretta di una sorgente di $^{109}$Cd (media 320 count), in base alla statistica accumulata sui fondi. La std raggiunge il minimo a partire da 7000 count.**